IN THE UNITED STATES PATENT AND TRADEMARK OFFICE

UTILITY PATENT APPLICATION

ELECTROSURGICAL JAW STRUCTURE FOR CONTROLLED ENERGY DELIVERY

BACKGROUND OF THE INVENTION

Field of the Invention

This invention relates to electrosurgical jaws and methods for delivering energy to tissue, and more particularly to an instrument working end for grasping tissue that self-modulates energy application to engaged tissues for sealing, welding or coagulating purposes.

Description of the Related Art

In various open and laparoscopic surgeries, it is necessary to coagulate, seal or weld tissues. One preferred means of tissue-sealing relies upon the application of electrical energy to captured tissue to cause thermal effects therein for sealing purposes. Various mono-polar and bi-polar radiofrequency (Rf) jaw structures have been developed for such purposes. In a typical bi-polar jaw arrangement, each jaw face comprises an electrode and Rf current flows across the captured tissue between the first and second polarity electrodes in the opposing jaws. While such bi-polar jaws can adequately seal or weld tissue volumes that have a small cross-section, such bi-polar instruments often are ineffective in sealing or welding many types of tissues, such as anatomic structures having walls with irregular or thick fibrous content, bundles of disparate anatomic structures, substantially thick anatomic structures, or tissues with thick fascia layers such as large diameter blood vessels.

Prior art Rf jaws that engage opposing sides of a tissue volume typically cannot cause uniform thermal effects in the tissue, whether the captured tissue is thin or substantially thick. As Rf energy density in tissue increases, the tissue surface becomes desiccated and resistant to additional ohmic heating. Localized tissue desiccation and charring can occur almost instantly as tissue impedance rises, which then can result in a non-uniform seal in the tissue. The typical prior art Rf jaws can cause a further undesirable effects by propagating Rf density laterally from the engaged tissue to cause unwanted collateral thermal damage.

What is needed is an instrument with a jaw structure that can apply Rf energy to tissue in new modalities: (i) to weld or seal tissue volumes that have substantial fascia layers or tissues that are non-uniform in hydration, density and collagenous content; (ii) to weld a targeted tissue region while substantially preventing thermal damage in regions lateral to the targeted tissue; and (iii) to weld a bundle of disparate anatomic structures.

SUMMARY OF THE INVENTION

The principal objective of the present invention is to provide an instrument and jaw structure that is capable of controllably applying energy to engaged tissue. As background, the biological mechanisms underlying tissue fusion by means of thermal effects are not fully understood. In general, the application of Rf energy to a captured tissue volume causes *ohmic* heating (alternatively described as *active* Rf heating herein) of the tissue to thereby at least partially denature proteins in the tissue. By ohmic heating, it is meant that the active Rf current flow within tissue between electrodes causes frictional or resistive heating of conductive compositions (e.g., water) in the tissue.

One objective of the invention is to denature tissue proteins, including collagen, into a proteinaceous amalgam that intermixes and fuses together as the proteins renature. As the treated region heals over time, the so-called weld is reabsorbed by the body's wound healing process. A more particular objective of the invention is to provide a system that (i) instantly and automatically modulates ohmic heating of tissue to maintain a selected temperature in the tissue, and (ii) to instantly and automatically modulate total energy application between active Rf heating (resulting from tissue's resistance to current flow therethrough) and conductive heating of tissue that results from heat conduction and radiation from resistively heated jaw components.

In general, the various jaw structures corresponding to the present invention all provide an Rf working end that is adapted to instantly and automatically modulate between active Rf heating of tissue and conductive heating of tissue by resistive jaw portions. Thus, the targeted tissue can be maintained at a selected temperature for a selected time interval without reliance of prior art "feedback" monitoring systems that measure impedance, temperature, voltage or a combination thereof.

In an exemplary embodiment, at least one jaw of the instrument defines a tissue-engagement plane that engages the targeted tissue. A cross-section of the jaw inwardly of the engagement plane illustrates that multiple electrically-conductive components comprise the invention for applying energy to tissue. Typically, the engagement plane defines a surface conductive portion (for tissue contact) that overlies a medial portion of a variably resistive material. An exemplary jaw further carries a core conductive material (or electrode) that is coupled to an Rf source and controller. Of particular interest, one embodiment has a variably resistive matrix that comprises a positive temperature coefficient (PTC) material having a resistance (i.e., impedance to electrical conduction therethrough) that changes as it increases in temperature. One type of PTC material is a ceramic that is engineered to exhibit a dramatically increasing resistance above a specific temperature of the material, sometimes referred to as a Curie point or a switching range.

In one embodiment, a jaw of the working end utilizes a medial variably resistive matrix that has a selected switching range, for example a 5°-20° C. range, which approximates a targeted temperature that is suitable for tissue welding. In operation, it can be understood that the engagement plane will apply active Rf energy to (or cause ohmic heating within) the engaged tissue until the point in time that the variably resistive matrix is heated to its selected switching range. When the tissue temperature thus elevates the temperature of the PTC material to the switching range, Rf current flow from the core conductive electrode through to the engagement surface will be terminated due to the temperature increase in tissue and the resistive matrix. This instant and automatic reduction of Rf energy application can be relied on to prevent any substantial dehydration of tissue proximate to the probe's engagement plane. By thus maintaining an optimal level of moisture around the engagement plane, the working end can more effectively apply energy to the tissue—and provide a weld thicker tissues with limited collateral thermal effects.

The working end of the probe corresponding to the invention further provides a suitable cross-section and mass for providing a substantial heat capacity. Thus, when the medial variably resistive matrix is elevated in temperature to its switching range, the matrix can effectively function as a resistive electrode to thereafter passively conduct thermal energy to the engaged tissue volume. Thus, in operation, the working end can automatically modulate the application of energy to tissue between *active* Rf heating and *passive* conductive heating of the targeted tissue to maintain the targeted temperature level.

In another preferred embodiment of the invention, the variably resistive matrix can be a silicone-based material that is flexible and compressible. Thus, the engagement surface of one or both jaws can flexibly engage tissue to maintain tissue contact as the tissue shrinks during the welding process. In a related embodiment, the variably resistive matrix can be an open-cell silicone-based material that is coupled to a fluid inflow source for delivering fluid to the engagement plane to facilitate welding of very thin tissue volumes.

The jaws of the invention can operate in mono-polar or bi-polar modalities, with the variably resistive matrix carried in either or both jaws of the working end.

DESCRIPTION OF THE DRAWINGS

Other objects and advantages of the present invention will be understood by reference to the following detailed description of the invention when considered in combination with the accompanying Figures, in which like reference numerals are used to identify like elements throughout this disclosure.

FIG. 1 is perspective view of a Type "A" working end of the invention showing first and second jaws carried at the end of an introducer.

FIG. 2 is a partial sectional view of a portion of the jaws of FIG. 1 taken along line 2-2 of FIG. 1 showing the active electrical energy delivery components corresponding to the invention.

FIG. 3 is a graph of the temperature vs. resistance profile of the positive temperature coefficient (PTC) matrix of the jaws of FIGS. 1-2.

FIG. 4 is a graph showing the temperature-resistance profile of the PTC matrix, the impedance of tissue and the combined resistance of the PTC matrix and tissue that is readable by feedback circuitry.

FIG. 5 is a sectional view of a Type "B" jaw structure that carries exposed first and second polarity electrodes and a variably resistive or PTC matrix in an engagement surface of a first jaw with an insulated second jaw, the first and second polarity electrodes adapted, in part, for bi-polar Rf energy application to engaged tissue.

FIG. 6 is a sectional view of an alternative Type "B" jaw structure similar to FIG. 5 that carries cooperating first and second polarity electrodes in both jaws, and a variably resistive matrix in a single jaw.

FIGS. 7A-7B are sectional views of another alternative Type "B" jaw structure similar to FIG. 6 that carry cooperating first and second polarity electrodes in both jaws, together with a variably resistive matrix in both jaws.

FIG. 8 is a sectional view of another Type "B" jaw structure similar to FIG. 6 that carry cooperating first and second polarity electrodes in both jaws, together with a variably resistive matrix in both jaws.

FIG. 9 is a sectional view of another Type "B" jaw structure similar to FIG. 8 that carries cooperating first and second polarity electrodes in each jaws, together with a variably resistive matrix in each jaw and an insulative outer layer.

FIG. 10 is a sectional view of another Type "B" jaw structure similar to FIG. 8 that has a first engagement plane with an exposed first polarity electrode and an interior variably resistive matrix and a second engagement plane with an exposed variably resistive matrix and an interior second polarity electrode, each jaw having an insulative outer layer.

FIG. 11 is a sectional view of yet another Type "B" jaw structure that has a first engagement plane with an exposed variably resistive matrix and an interior first polarity electrode, a second engagement plane with an exposed second polarity electrode.

FIG. 12 is a sectional view of another Type "B" jaw structure that has a first and second engagement planes with an exposed variably resistive matrix and interior first and second polarity electrode, respectively.

FIG. 13 is a perspective view of a Type "C" working end in accordance with the invention that has a jaw engagement plane with a variably resistive (PTC) material that is flexible or compressible.

FIG. 14 is a sectional view of the jaws of FIG. 13 showing the variably resistive (PTC) material and exposed electrode surfaces.

FIG. 15 is a sectional view of a Type "D" jaw in accordance with the invention that has a jaw engagement plane comprising a surface of an open-cell, compressible variably resistive (PTC) material together with a fluid source is coupled thereto for delivering a fluid to the engagement plane.

FIG. 16 is a cut-away view of a portion of a Type "E" jaw that carries opposing polarity electrodes with spaced apart volumes of a thermally-sensitive resistive matrix.

FIG. 17 is a sectional view of a portion of a Type "F" jaw structure that carries variably resistive layer that is pressure sensitive.

DETAILED DESCRIPTION OF THE INVENTION

1. Type "A" working end for tissue sealing. An exemplary Type "A" working end **100** of a surgical grasping instrument is illustrated in FIGS. 1-2 that is adapted for energy delivery for sealing or welding tissue. The working end **100** is carried at the distal end of an introducer portion **102** that can be rigid or flexible in any suitable diameter. For example, the introducer portion **102** can have a diameter ranging from about 3 mm. to 5 mm. (or larger) for use in endoscopic surgical procedures. The introducer portion extends along axis **107** from its proximal end that is connected to a handle (not shown). The working end has first (lower) jaw **105A** and second (upper) jaw **105B** that are coupled to the distal end **108** of the introducer portion **102**. The jaws may both be moveable or a single jaw may move to provide an open position and a closed position wherein the jaws approximate toward axis **107**. The opening-closing mechanism can be any type known in the art. For example, a reciprocatable cam-type member **109** can slide over the jaws **105A** and **105B** to engage the outer surfaces of the jaws, and can be the type of mechanism disclosed in co-pending U.S. Patent Application Ser. No. 09/792,825 filed February 24, 2001 (Docket No. SRX-006) titled *Electrosurgical Working End for Transecting and Sealing Tissue*, which is incorporated herein by reference.

In the exemplary embodiment of FIG. 2, the first (lower) jaw **105A** has a tissue-engaging surface or engagement plane **125** that contacts and delivers energy to a targeted tissue. The jaws can have any suitable length with teeth or serrations in any location for gripping tissue, and is shown in FIG. 2 with such serrations **127** along the outboard portions of the jaws thus leaving the engagement plane **125** inward of the serrations. In the embodiments described below, the engagement plane **125** generally is shown with a non-serrated surface for clarity of explanation, but the engagement plane **125** itself can be any non-smooth gripping surface.

In the exemplary embodiment of FIG. 2, the engagement surface or plane **125** that delivers energy to tissue extends along an axial length of only the lower jaw **105A**, and the tissue-contacting surface **130** of the second jaw **105B** is passive and comprises an insulated material **131** or has an insulated surface layer. As will be described below, alternative embodiments corresponding to the invention provide both the first and second jaws with electrically "active" components.

The sectional view of FIG. 2 more particularly illustrates the individual electrically relevant components within the body of the lower jaw for controllably delivering energy to tissue for sealing or welding purposes. The engagement surface **125** of jaw **105A** has a conductive surface material indicated at **135** that is both electrically conductive and thermally conductive. For example, the conductive surface layer **135** can be a thin film deposit of any suitable material known in the art (e.g., gold, platinum, palladium, silver, stainless steel, etc.) having any suitable thickness dimension $\mathbf{d_1}$, for example, ranging from about of 0.0001" to 0.020". Alternatively, the conductive layer **135** can comprise a machined or cast metal having a more substantial thickness that is conductively bonded to the interior layers described next.

As can be seen in FIG. 2, the jaw **105A** has a medial (second) material or matrix **140** that is *variably resistive* (alternatively called *variably conductive* herein) and carried inwardly of the surface conductive material **135**. Further, the body of jaw **105A** carries a (third) interior conductive material or electrode **145** at its core. The medial conductive layer **140** thus is intermediate the engagement plane **125** and the interior conductive material **145**. The third conductive material or electrode **145** is coupled by an electrical lead to a remote voltage (Rf) source **150** and optional controller **155**. The medial variably resistive matrix **140** can have any suitable cross-sectional dimensions, indicated generally at $\mathbf{d_2}$ and

d_3, and preferably such a cross-section comprises a significant fractional volume of the jaw body to provide a thermal mass for optimizing passive conduction of heat to tissue as will be described below.

It can be easily understood from FIG. 2 that the core conductive material **145** is coupled to, or immediately adjacent to, the medial variably resistive material **140** for conducting electrical energy from the interior conductor to engaged tissue through matrix **140**. In FIG. 2, it can be seen that the first, second and third components (indicated at **135**, **140**, **145**) are carried in a structural body component **148** of the jaw **105A** that can be any suitable metal with an insulative coating or any other rigid body that can accommodate loads on the jaw as it engages and compresses tissue.

Of particular interest, still referring to FIG. 2, the medial variable conductive matrix **140** comprises a polymeric material having a temperature-dependent resistance. Such materials are typically known in the art as polymer-based temperature coefficient materials, and sometimes specifically described as thermally-sensitive resistors or thermistors that exhibit very large changes in resistance with a small change of body temperature. This change of resistance with a change in temperature can result in a positive coefficient of resistance where the resistance increases with an increase in temperature (PTC or positive temperature coefficient material). The scope of the invention also includes a medial variably conductive matrix **140** of a negative temperature coefficient (NTC) material wherein its resistance decreases with an increase in temperature.

In one preferred embodiment, the PTC matrix **140** is a ceramic layer that can be engineered to exhibit unique resistance vs. temperature characteristics and can maintain a very low base resistance over a wide temperature range, with a dramatically increasing resistance above a specific temperature of the material (sometimes referred to as a Curie point or *switching range*; see FIG. 3). One aspect of the invention relates to fabrication of the medial PTC matrix **140** to have a selected switching range between a first temperature (T_1) and a second temperature (T_2) that approximates the targeted tissue temperature in the contemplated tissue sealing or welding objective. The selected switching range, for example, can be any substantially narrow 1°-10° C. range that is determined to be optimal for tissue sealing or welding (e.g., any 5° C. range between about 65°-200° C.). A more preferred switching range can fall within the larger range of about 80°-100° C.

In operation, it can be understood that the delivery of Rf energy to the interior conductor **145** will be conducted through the variably conductive matrix **140** and the engagement plane **125** to thereby apply Rf energy (or active ohmic heating) to tissue engaged between the jaws **105A** and **105B** (see FIG. 2). After the engaged tissue is elevated in temperature by such active Rf heating, the lower jaw's conductive surface layer **135** and the medial conductive layer **140** will be elevated to the selected switching range. Thereafter, the mass of the body will be modulated in temperature, similar to the engaged tissue, at or about the selected switching range. Thereafter, the jaws body will conduct or radiate thermal effects to the engaged tissue.

In other words, the critical increase in temperature of the variably resistive matrix **140** is typically caused by the transient high temperature of tissue that is caused by active Rf heating of the tissue. In turn, heat is conducted back through the layer of the first conductive material **135** to medial matrix **140**. A suitable variably resistive PTC material can be fabricated from high purity semi-conducting ceramics, for example, based on complex titanate chemical compositions (e.g., $BaTiO_3$, $SrTiO_3$, etc.). The specific resistance-temperature characteristics of the material can be designed by the addition of dopants and/or unique materials processing, such as high pressure forming techniques and precision sintering. Suitable variably resistive or PTC materials are manufactured by a number of sources, and can be obtained, for example from Western Electronic Components Corp., 1250-A Avenida Acaso, Camarillo, CA 93012. Another manner of fabricating the medial conductive material **140** is to use a commercially available epoxy that is doped with a type of carbon. In fabricating a substantially thin medial conductive layer **140** in this manner, it is preferable to use a carbon type that has single molecular bonds. It is less preferable to use a carbon type with double bonds which has the potential of breaking down when used in thin layers, thus creating the potential of an electrical short circuit between conductive portions **145** and **135**.

As can be seen in FIG. 2, the core conductive material or electrode **145** is operatively connected to the voltage (Rf) source **150** by a first electrical lead **156a** that defines a first polarity of the Rf source. In this preferred embodiment, the conductive engagement surface **135** is coupled to a second electrical lead **156b** that defines a second or opposing

polarity of the Rf source **150**. A ground pad indicated at **158** in FIG. 2 first lead **156a** to accomplish a preferred method of the invention, as will be described below.

The manner of utilizing the working end **100** of FIG. 2 to perform a method of the invention can be understood as engaging and compressing tissue between the first and second jaws **105A** and **105B** and thereafter applying active Rf energy to the tissue to maintain a selected temperature for a selected time interval. For example, the instrument is provided with a working end that carries a medial variably conductor matrix **140** (see FIG. 2) that has a switching range at or about 90° C. at which its resistance increases greater than about 5% (and can be as much as 1,000,000% or more) above its low base resistively with a change in temperature of about 5° C. or less (see FIG. 3).

With the jaws in the closed position and the engagement plane **125** engaging tissue, the operator actuates a switch **164** that delivers Rf energy from the voltage (Rf) source **150** to the interior conductor **145**. At ambient tissue temperature, the low base resistance of the medial conductive matrix **140** allows unimpeded Rf current flow from the voltage source **150** through the engagement surface **125** (and conductor layer **135**) and tissue to return electrical lead **156a** that is coupled to ground pad **158**. It can be understood that the engaged tissue initially will have a substantially uniform impedance to electrical current flow, which will increase substantially in proximity to engagement surface **125** as the engaged tissue loses moisture due to the active Rf delivery.

Following an arbitrary time interval, the impedance of tissue proximate to engagement surface **125** typically will be elevated, and the higher tissue temperature will instantly conduct heat to the medial PTC matrix **140**. In turn, the medial PTC layer **140** will reach its switching range and terminate Rf current flow from the core conductor **145** to the engagement surface **125**. Such automatic reduction of active Rf energy application will prevent any substantial dehydration of tissue proximate to the engagement plane **125**. By thus maintaining the desired level of moisture in tissue proximate to the engagement plane **125**, the working end can more effectively apply energy to the tissue. Such energy application can extend through thick engaged tissue volumes while causing very limited collateral thermal effects. Thereafter, as the temperature of tissue proximate to engagement surface **125** falls by thermal relaxation and the lack of an Rf energy density, the temperature of the medial conductive matrix **140** will thus fall below the threshold of the

selected switching range. This effect, in turn, will cause Rf current to again flow through the assembly of conductive layers **145**, **140** and **135** to the engaged tissue to again increase the tissue temperature by active Rf heating. By the above-described mechanisms of causing the medial variably resistive matrix **140** to hover about its selected switching range, the actual Rf energy applied to the engaged tissue can be precisely modulated to maintain the desired temperature in the tissue.

Of particular interest, in one embodiment, the polymer matrix that comprises the medial conductor portion **140B** is doped with materials to resistively heat the matrix as Rf energy flow therethrough is reduced. Thus, the thermal mass of the jaws which are elevated in temperature can deliver energy to the engaged tissue by means of greater *passive* conductive heating—at the same time Rf energy delivery causes lesser *active* tissue heating. This balance of active Rf heating and *passive* conductive (or radiative) heating can maintain the targeted temperature for any selected time interval.

In summary, one method of the invention comprises the delivery of Rf energy from a voltage source **150** to a conductive jaw surface **135** through a thermally-sensitive resistor material **140** wherein the resistor material has a selected switching range that approximates a targeted temperature for tissue sealing or welding. In operation, the working end *automatically* modulates active Rf energy density in the tissue as the temperature of the engaged tissue conducts heat back to the thermally-sensitive resistor material **140** to cause its temperature to reach the selected switching range. In this range, the Rf current flow will be reduced, with the result being that the tissue temperature can be maintained in the selected range without the need for thermocouples or any other form of feedback circuitry mechanisms to modulate Rf power from the source. Most important, it is believed that this method of the invention will allow for immediate modulation of *actual* Rf energy application along the entire length of the jaws, which is to be contrasted with prior art instruments that utilize a temperature sensor and feedback circuitry. Such sensors or thermocouples measure temperature only at a single location in the jaws, which typically will not be optimal for energy delivery over the length of the jaws. Such temperature sensors also suffer from a time lag. Further, such temperature sensors provide only an indirect reading of actual tissue temperature—since a typical sensor can only measure the temperature of the electrode.

Another method of the invention comprises providing the working end with a suitable cross-section of variably resistive matrix **140** so that when it is elevated in temperature to a selected level, the conductive matrix **140** effectively functions as a resistive electrode to passively conduct thermal energy to engaged tissue. Thus, in operation, the jaws can automatically modulate the application of energy to tissue between *active* Rf heating and *passive* conductive heating of the targeted tissue at a targeted temperature level.

FIG. 3 illustrates another aspect of the method of the invention that relates to the Rf source **150** and controller **155**. A typical commercially available radiofrequency generator has feedback circuitry mechanisms that control power levels depending on the feedback of impedance levels of the engaged tissue. FIG. 3 is a graph relating to the probe of present invention that shows: (i) the temperature-resistance profile of the targeted tissue, (ii) the temperature-resistance profile of the PTC conductive matrix **140** of the probe, and (iii) the combined temperature-resistance profile of engaged tissue and the PTC conductive matrix. In operation, the Rf source **150** and controller **155** can read the combined impedance of the engaged tissue and the PTC conductive layer which will thus allow the use of the instrument with any typical Rf source without interference with feedback circuitry components.

2. <u>Type "B" jaw structures</u>. A series of exemplary Type "B" jaw structures **200a-200i** corresponding to the invention are illustrated in FIGS. 5-12. Each set of paired first and second jaws, **205A** and **205B**, are shown in sectional view to illustrate more specifically how a thermally-dependent variably resistive layer (indicated at **240** in FIGS. 5-12) can be configured to cooperate with at least one electrode to (i) apply *active* Rf energy to tissue engaged between the jaws while the temperature of the variably resistive matrix is below its switching range, and (ii) to cause modulation of both *active* and *passive* heating when the variably resistive matrix hovers about its switching range.

FIG. 5 illustrates working end **200a** wherein lower (first) jaw **205A** defines an engagement plane **225A** that contacts tissue. In this embodiment, the upper (second) jaw **205B** defines a tissue-contacting plane **225B** that is a surface of an insulator material **226**. The structural components of the jaws indicated at **228a** and **228b**, if required for strength, are of an insulated material or separated from the electrical body components by an insulative layer. Now turning to the active components of the working end **200a**, the thermally-dependent resistive layer **240** is exposed to the engagement

plane **225A** at regions **244a** and **244a'**. In this embodiment of FIG. 5, the thermally-dependent resistive layer **240** is intermediate to the opposing polarity electrodes **245A** and **245B** as defined by the circuitry coupled to voltage source **150** (see FIG. 1). For convenience, the electrodes are indicated throughout this disclosure as positive (+) and (-) polarities at a particular point in time. The electrodes **245A** and **245B** have surface portions **247a** and **247b** exposed in the engagement plane of the lower jaw. It should be appreciated that the size and shape of structural body components **228a-228b** can be varied, and may not be required at all. For example, a jaw as depicted in FIG. 5 can use a substantially strong metal for electrode **245A** which can comprise the structural body component of the jaw, with or without a thin insulative coating outside the engagement plane **225A**. For clarity of explanation, the gripping elements that are typically used in the jaw surface are not shown.

It can be understood from FIG. 5 that active heating of the targeted tissue **tt** (phantom view) will occur generally from current flow between first polarity electrode **245A** and second polarity electrode **245B** (see arrows **A**). Such current flow also can cooperate with a separate but optional "ground-pad" used as a return electrode (cf. FIG. 2). It can be further understood that the elevation of the tissue temperature of the medial PTC matrix **240** will then modulate energy application between (i) active Rf heating, and (ii) passive or conductive heating. For this reason, in this embodiment as the others described next, the medial PTC matrix **240** preferably comprises a substantial portion of the jaw body for retaining heat for such conductive heating.

FIG. 6 illustrates working end **200b** with the lower (first) jaw **205A** defining engagement plane **225A** and the upper (second) jaw **205B** defining engagement plane **225B**. This embodiment is very similar to the embodiment of FIG. 5, except that the upper engagement plane **225B** comprises a surface of an active conductive body indicated at **245B'** that is coupled to a voltage source. The electrode **245B'** has a polarity common with electrode **245B**, or alternatively can have a polarity common with electrode **245A** (not shown). It can be understood from FIG. 6 that active heating of tissue engaged between the jaws will occur as current flows between second polarity electrodes **245B-245B'** and the first polarity electrode **245A**. The working can also cooperate with a separate "ground-pad" that functions as a return electrode (not shown). As described previously, the elevation of the temperature of medial PTC matrix **240** again will

modulate energy application between (i) active Rf heating, and (ii) passive or conductive heating about its selected switching range. It should be appreciated that jaws **205A** and **205B** as depicted FIG. 6 with opposing polarity electrodes in direct opposition to one another can be provided with further means for preventing the electrodes from contacting each other as the jaws are pressed together. The perimeter of the jaws or the jaw's engagement planes can carry surface elements indicated at **249** (phantom view) that prevent contact of the opposing polarity electrodes. For example, the active electrode surfaces can be slightly recessed relative to such elements **249**.

FIGS. 7A-7B illustrates working ends **200c** and **200d** that are similar and show optional configurations of jaws **205A** and **205B** that define engagement planes **225A** and **225B**, respectively. Each jaw in the embodiments of FIGS. 7A-7B carry opposing polarity electrodes with an intermediate layer of a thermally-dependent resistive matrix **240**. In FIG. 7A, electrodes **245A** and **245B** have exposed surfaces **244a** and **244b** in the lower jaw's engagement plane **225A**. Similarly, electrodes **245A'** and **245B'** have exposed surfaces **244a'** and **244b'** in engagement plane **225B** of the upper jaw. The electrode arrangement of FIG. 7B differs only in the spatial location of the opposing polarity electrode surfaces. In use, it can be understood how Rf active heating of engaged tissue will occur as current flows between first polarity electrodes in each jaw—and between the jaws. Again, the elevation of the temperature of medial PTC matrix **240** will modulate energy application between active Rf heating and passive heating at the selected switching range.

In another similar electrode arrangement (not shown), a plurality of common polarity electrodes can be exposed in an engagement surface with a phase shift in the voltage delivered to such electrodes as provided by the voltage source or sources. Such phase shift electrodes can cooperate with a return electrode in either jaw's engagement surface and/or a ground pad. The paired jaw's engagement surfaces also can be configured with mirror image electrodes, which can be phase shift electrodes which can reduce capacitive coupling among such electrode arrangements. In general , such phase shift features can be combined with any of the working ends of FIGS. 6-12.

FIG. 8 illustrates a sectional view of another embodiment of working end **200e** with jaws **205A** and **205B** that define engagement planes **225A** and **225B**, respectively. In this embodiment, each jaw's engagement plane carries at least two spaced apart electrode surfaces with opposing polarities. For example, lower jaw **205A** carries first polarity

electrodes **245A** and second polarity electrode **245B** (collectively). Between the electrodes **245A** and **245B** is an intermediate thermally-dependent resistive matrix **240**. The upper jaw **205B** carries first polarity electrodes **245A'** (collectively) and second polarity electrode **245B'** with thermally-dependent resistive matrix **240** therebetween. Such a jaw configuration will modulate the application of energy to tissue as described previously. FIG. 9 illustrates a sectional view of the jaws of working end **200f** which is similar to FIG. 8 except that the jaws have an outer perimeter of a insulator material indicated at **251**.

FIGS. 10 and 11 illustrate sectional views of other embodiments of working end **200g** and **200h** with jaws **205A** and **205B** that define engagement planes **225A** and **225B**, respectively. In these embodiments, one jaw's engagement plane has an active exposed electrode surface indicated at **245B**, while the opposing jaw carries an opposing polarity electrode **245A** at its interior with a thermally-dependent resistive matrix **240** at that jaw's engagement surface. The jaw assembly of FIG. 10 further shows an optional insulative layer **251** about the exterior of the jaws. The resistive matrix comprises a substantial portion of the jaw's mass, and thus is adapted to modulate the application of energy to tissue as described previously.

FIG. 12 illustrates another sectional view of a jaw assembly **200i** corresponding to the invention with jaws **205A** and **205B** that define engagement planes **225A** and **225B**, respectively. In this embodiment, each jaw's engagement plane **225A** and **225B** comprises a surface of a thermally-dependent resistive matrix **240** with neither of the opposing polarity electrodes **245A** and **245B** being exposed for tissue contact.

3. <u>Type "C" jaw structure for sealing tissue</u>. An exemplary Type "C" jaw structure **300** carried by introducer **310** corresponding to the invention is illustrated in FIGS. 13-14. The Type "C" system differs in that it utilizes a different form of thermally-dependent resistive layer (indicated at **340** in FIGS. 13-14) that is an elastomer, for example a silicon-based sponge-type material that can be resilient or compressible. More in particular, FIG. 13 illustrates working end **300** with lower (first) jaw **305A** defining engagement plane **325A** that contacts tissue. The upper (second) jaw **305B** defines a tissue-contacting plane **325B** that is a surface of an insulator material **326**, but it can also carry electrically conductive components as generally depicted in FIGS. 6-12. The jaw structure of FIG. 13 further shows that it is

configured with a central channel or slot **332** that is adapted to accommodate a reciprocatable tissue-cutting member **333** for transecting sealed tissue. Such a moveable cutting member **333** is actuated from the instrument handle (not shown) as is known in the art. The cutting member **333** can be a sharp blade or an Rf cutting electrode that is independently coupled to a high voltage Rf source. This embodiment shows tissue-gripping serrations **334** along an inner portion of the jaws, but any location is possible. It should be appreciated that electrode components and thermally-dependent resistive components of the invention can be adapted for any jaws, or left-side and right-side jaw portions, in (i) conventional jaws for tissue sealing or (ii) combination jaws for sealing-transecting instruments. In the embodiment depicted in FIGS. 13-14, the structural body of the jaws **328a** and **328b** again are preferably of an insulated material or separated from the electrically-connected materials by an insulative layer.

A principal purpose for providing a flexible variably conductive matrix **340** is to dynamically adjust the pressure of the engagement plane **325A** against the tissue volume that is compressed between the jaws. It is believed useful to provide a dynamic engagement plane since tissue may shrink during a sealing procedure. The repose or untensioned shape of the variably conductive matrix **340** is shown as being convex, but also can be flat in cross-section or it can have a variety of different geometries or radii of curvature.

Of particular interest, the objective of a resilient jaw surface has resulted in the development of an assembly of materials that can provide a *flexible* and *resilient* engagement plane **325A** at the surface of a resilient variably resistive material **340**. In the embodiment of FIGS. 13-14, it can be seen that the conductive portion or electrode **345** is a thin metallic layer or member bonded to the variably resistive matrix **340**, which defines exposed surfaces **347** in the engagement plane **325A**. The conductive electrode **325** again coupled to electrical source **150** and controller **155**, as described previously.

Of particular interest, the variably resistive matrix **340** comprises a silicone material that can function as a PTC-type resistive matrix in the same manner as the above-described ceramic materials. More in particular, one embodiment of the variably resistive matrix **340** can be fabricated from a medical grade silicone that is doped with a selected volume of conductive particles, e.g., carbon or graphite particles. By weight, the ratio of carbon to silicone can range from about

90/10 to about 30/70 to provide various selected switching ranges wherein the inventive composition then functions as a positive temperature coefficient (PTC) material. More preferably, the matrix is form about 40% to 80% carbon with the balance being silicone. As described previously, carbon types having single molecular bond are preferred. One preferred composition has been developed to provide a switching range of about 75° C. to 90° C. has about 50%-60% carbon with the balance being silicone. The variably resistive matrix **340** can have any suitable thickness dimension indicated at $\mathbf{d_4}$, ranging from about 0.01" to 0.25" depending on the cross-section of the jaws.

The electrode **345** that is exposed in engagement plane **325A** can be a substantially thin rigid metal, flexible foil, or a substantially flexible thin metallic coating. Such a thin flexible coating can comprise any suitable thin-film deposition, such as gold, platinum, silver, palladium, tin, titanium, tantalum, copper or combinations or alloys of such metals, or varied layers of such materials. A preferred manner of depositing a metallic coating on the polymer element comprises an electroless plating process known in the art, such as provided by Micro Plating, Inc., 8110 Hawthorne Dr., Erie, PA 16509-4654. The thickness of the metallic coating can range between about 0.0001" to .005". Other similar electroplating or sputtering processes known in the art can be used to create a thin film coating.

In operation, the working end of will function as described in the Types "A" and "B" embodiments. The elevation of the tissue temperature will conduct heat directly to the PTC matrix **340** and then will modulate energy application between (i) active Rf heating, and (ii) passive or conductive heating. In addition, the resiliency of the PTC matrix **340** will maintain substantially uniform pressure against the tissue even as the tissue dehydrates or shrinks.

While the sectional view of the jaws **305A** and **305B** of FIGS. 13-14 depict a preferred embodiment, it should be appreciated that jaws **305A** and **305B** can use a compressible-resilient PTC matrix **340** in *any* of the electrode-PTC matrix configurations shown in FIGS. 5-12, all of which fall within the scope of the invention.

5. <u>Type "D" jaw structure for sealing tissue</u>. An exemplary working end of a Type "D" probe **400** is shown in FIG. 15 that again is adapted for energy delivery to an engaged tissue volume for sealing or welding purposes. The jaws **405A** and **405B** define engagement planes **425A** and **425B** as described previously that engage tissue from opposing sides. The lower (first) jaw **405A** again carries variably resistive portion **440** that is a resilient sponge-type material, e.g.,

a silicone-based material, that is very similar to that described in the Type "C" embodiment above. The PTC matrix **440** in this embodiment comprises an *open cell* structure of the silicon polymer or other sponge polymer. More in particular, FIG. 15 illustrates working end **400** with lower (first) jaw **405A** defining engagement plane **425A** that contacts tissue. The tissue-contacting plane of upper jaw (not shown) can be the same as illustrated in FIG. 14 and comprise an insulator material. Alternatively, the upper jaw can carry electrically conductive body components that match the lower jaw. The jaw structure again is shown in FIG. 15 with a central channel **432** for accommodating a reciprocatable cutting member.

Of particular interest, in the embodiment of FIG. 15, the system is adapted to deliver saline flow from fluid source **460** directly through the *open cell* structure of the silicon-based PTC conductive layer **460**. Such an open cell silicone can be provided adding foaming agents to the silicone during its forming into the shape required for any particular working end. The silicone has a conductive material added to matrix as described above, such as carbon. In this embodiment, an exposed electrode surface **445** comprises an elongate conductive element that exposes portions of the compressible PTC conductive portion **440**. Alternatively, the electrode surfaces can be a thin microporous metallic coating, of the types described previously. The electrode **445** is shown as cooperating with a ground pad **458**, although any of the electrode and resistive matrix arrangements of FIGS. 5-12 fall within the scope of the invention.

In a method of using the jaw structure of FIG. 15, the system can apply saline solution through pores **462** in the open cell matrix **440** that are exposed in the engagement plane **425A** that engages tissue. The method of the invention provides for the infusion of saline during an interval of energy application to engaged tissues to enhance both active Rf heating and conductive heating as the jaws maintain tissue temperature at the selected switching range of the PTC matrix **440**. In another aspect of the invention, the compressibility of the silicone-based medial conductive portion **440** can alter the volume and flow of saline within the open cell silicone PTC portion **440**. Since the saline is conductive, it functions as a conductor within the cell voids of the medial resistive matrix **440**, and plays the exact role as the carbon doping does within the walls of cells that make up the silicone. Thus, the extent of expansion or compression of the silicone medial conductive portion **440** alters its resistivity, wherein the conductive doping of the material remains static. It can be understood that a compression of PTC matrix **440** can collapse the cells or pores **462** which in turn will restrict fluid

flow. Thus, the system can be designed with (i) selected conductive doping of silicone PTC matrix **440** and (ii) selected conductivity of the saline solution to optimize the temperature coefficient of the material under different compressed and uncompressed conditions for any particular tissue sealing procedure. The sponge-type variably resistive body portion **440** can be designed to be a positive or negative temperature coefficient material (defined above) as the material expands to a repose shape after being compressed. The resilient engagement surface **425A** can naturally expand to remain in substantial contact with the tissue surface as the tissue is sealed and dehydrates and shrinks. At the same time, the cell structure of the medial conductive portion **440** will tend to open to thereby increase fluid flow the engagement plane, which would be desirable to maintain active and passive conductive heating of the tissue. Also at the same time, the selected temperature coefficient of the silicone PTC matrix **440** in combination with the saline volume therein can insure that active Rf heating is modulated as exactly described in the Types "A" and "B" embodiments above with any selected switching range. It is believed that the use of saline inflow will be most useful in welding substantially thin tissue volumes that could otherwise desiccate rapidly during active Rf energy delivery. Thus, this effect can be used to design into the working end certain PTC characteristics to cause the working end to perform in an optimal manner.

It should be appreciated that the scope of the invention includes the use of an open cell elastomer such as silicone to make both a *temperature-sensitive* variable resistive matrix and a form of *pressure-sensitive* resistive matrix. As described above, the matrix under compression will collapse pores in the matrix thereby making a conductively-doped elastomer more conductive. In effect, such a matrix can be described as a pressure-sensitive matrix or a combination temperature-sensitive and pressure-sensitive variably resistive matrix. Further, an open cell elastomer that is *not* conductively-doped can function as a variably resistive matrix in combination with conductive fluid. When under compression, the conductive characteristics of the matrix would be lessened due to the outflow of the conductive fluids. Thus, it can be seen that the variably resistive matrix can be designed in a variety of manners to accomplish various Rf energy delivery objectives, all of which fall within the scope of the invention.

6. <u>Type "E" jaw structure for sealing tissue</u>. An exemplary working end of a Type "E" probe **500** is described with reference to FIG. 16. The Type "E" jaw of FIG. 16 is adapted for controlled energy delivery and is provided with

additional features that allows the variably resistive matrix to function optimally in working ends with a small cross-section. The objective to the Type "E" embodiment is to greatly reduce capacitive losses in operation.

In FIG. 16, it can be seen that working end **500** has first jaw **505A** (second jaw **505B** not shown) that extends along axis **507** with the lower jaw alone carrying the active energy delivery components of the invention, although both jaws could have such active components. The lower jaw **505A** defines an engagement plane **525** with cooperating jaw body components that comprise opposing polarity conductive portions **535A** and **535B** with intermediate elements of a thermally-sensitive resistive matrix indicated at **540**. For clarity of explanation, the opposing polarity portions **535A** and **535B** are indicated as positive (+) and negative (-). The resistive matrix **540** can be any of the types described above and preferably is a rigid ceramic-type positive temperature coefficient (PTC) material (see FIG. 3).

In the Types "A" and "B" working ends of FIGS. 5-9 above, the active jaw carried opposing polarity conductive portions and with an intermediate layer of a variably resistive matrix—which is similar to the Type "E" working end **500**. As can be seen, for example in FIG. 6, the transverse dimension **td** across the variably resistive matrix between the opposing polarity electrodes can be substantially small due to the thin layers of material. The direction of current flow is indicated at **A** in FIG. 6 which is generally transverse to the axis **107** of the jaws (see FIG. 6).

The Type "E" embodiment of FIG. 16 provides an improved manner of arranging the conductive components of a working end to reduce capacitive coupling of opposing polarity conductive portions **535A** and **535B** across the variably resistive matrix **540**. More in particular, the Type "E" positions elements of the opposing polarity conductive portions **535A** and **535B** is such a manner to induce a selected *directional* current flow through the resistive matrix **540**, with the preferred direction being any elongated dimension within the jaw structure—typically not a transverse direction across the jaw. The typical preferred direction for inducing current flow is an axial direction in relation to axis **507**, with such current flow through matrix indicated at arrow **AA** in FIG. 16. It should be appreciated that the arrangement of jaw components in FIG. 16 can be provided in jaws that have a channel for receiving a reciprocating blade member as shown in FIGS. 13-14, or combined with any of the electrode configurations of FIGS. 5-12.

In one Type "E" embodiment corresponding to the invention is depicted in FIG. 16, the jaw has at least one volume of resistive matrix **540** that comprises an interior body portion of the jaw (similar to the Type "A" embodiment of FIG. 2). Each said volume of resistive matrix **540** has first and second ends **542a** and **542b** that contact the respective opposing polarity conductive portions **535A** and **535B**. The axial dimension **ad** between the first and second ends **542a** and **542b** can be any suitable dimension and can be substantially greater than the thickness dimension **td** of the resistive matrix **540**. The conductive portions **535A** and **535B** each have a projecting leg portion (**546a** and **546b**) that contact the first and second ends **542a** and **542b** of the resistive matrix **540**. As can be seen in FIG. 16, the body portions of the resistive matrix **540**, except for the first and second ends **542a-542b** thereof, are surrounded by electrically insulative layers indicated at **548**. The opposing polarity conductive portions **535A** and **535B** are coupled to a voltage source. Thus, it can be understood how current is induced to flow in the direction of arrow **AA** through the matrix **540** to reduce capacitive coupling across the resistive matrix **540**. The insulative layer **548** can be any type of material or layer, for example, a thin layer of a titanium oxide ceramic-type material. In another embodiment (not shown), the insulative layer **548** can be an air space. In yet another embodiment, the insulative layer **548** can any combination of air spaces and any other insulative material.

7. <u>Type "F" jaw structure for sealing tissue</u>. FIG. 17 illustrates one jaw of a Type "F" system **600** corresponding to the invention. The jaw structure again is adapted for controlled energy delivery to tissue utilizing a variably resistive matrix **640**. The lower jaw **605A** defines an engagement plane or surface **625** for contacting tissue that overlies two variably resistive body portions indicated at **630** and **640**. The core of the jaw **605A** again carries a conductive body portion indicated at **645** that is coupled to a voltage source as described previously. The interior variably resistive matrix **640** is a thermally sensitive material as described in the Types "A" and "B" embodiments, for example, a PTC matrix of a ceramic material. An optional structural body of the jaws is indicated at **647** which is insulated from the above-described electrically active components.

Of particular interest, the exterior variably resistive matrix **630** is of a pressure-sensitive resistive material that is carried across the engagement plane **625**. In one embodiment, such a variably resistive layer **630** can be substantially thin and fabricated of a material described as a "pressure variable resistor ink" and is more specifically identified as

Product No. CMI 118-44 available from Creative Materials Inc., 141 Middlesex Rd., Tyngsboro, MA 01879. The resistance vs. pressure characteristics of the pressure-sensitive resistive matrix **630** can be adjusted by blending the above-described material with Product No. CMI 117-34 that is available from the same source.

In operation, it can be understood that any pressure against the pressure-sensitive resistive layer **630** will locally decrease its resistance to current flow therethrough. As the jaws are closed, the engagement plane **625** of the lower jaw **605A** will be pressed against the tissue. Due to the potential pressure vs. resistivity characteristics of the resistive layer **630**, the layer can be designed so that Rf current will only flow through localized portions of the engagement plane **625** where the pressure-sensitive resistive layer **830** is under substantial pressure, which in turn locally lowers the resistance of a portion of the surface layer. Further, the interior thermally-sensitive variable sensitive resistive layer **640** will modulate Rf flow as previously described to maintain a targeted tissue temperature.

It should be appreciated that the scope of the invention and its method of use of the includes the use of jaw working surface similar to that of FIG. 17 that does not carry an interior body portion of the thermally-sensitive resistive matrix indicated at **640**. In other words, the jaw can rely only on the pressure-sensitive resistive layer **630** about the engagement plane **625** to locally apply energy to captures tissue volumes.

In another embodiment (not shown), either one both jaws can have an elongate core of the substantially resistive material in addition to the core electrode and a variably resistive matrix of any type described above. The resistive material has a *fixed* resistance and is adapted to pre-heat the jaw, its engagement plane and the engaged tissue as a means of pre-conditioning the tissue to attain a certain selected impedance. Such a system will be useful when the engagement plane is large in dimension. A thermally conductive, but electrically insulative, layer can be disposed intermediate the fixed resistance material and a conductive (electrode) layer. The conductive layer is coupled in series with the fixed resistance material to the remote voltage source. The variably resistive matrix is disposed between the engagement plane and the conductive (electrode) layer—as described in any of the Types “A” and “B” embodiments above.

Those skilled in the art will appreciate that the exemplary systems, combinations and descriptions are merely illustrative of the invention as a whole, and that variations of components, dimensions, and compositions described above may be made within the spirit and scope of the invention. Specific characteristics and features of the invention and its

method are described in relation to some figures and not in others, and this is for convenience only. While the principles of the invention have been made clear in the exemplary descriptions and combinations, it will be obvious to those skilled in the art that modifications may be utilized in the practice of the invention, and otherwise, which are particularly adapted to specific environments and operative requirements without departing from the principles of the invention. The appended claims are intended to cover and embrace any and all such modifications, with the limits only of the true purview, spirit and scope of the invention.